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ON THE DEVELOPMENT OF LIFE PREDICTION METHODOLOGIES FOR THE FAILURE OF HUMAN TEETH

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Abstract

Human dentin is known to be susceptible to failure under cyclic loading. Surprisingly, there are few reports that quantify the effect of such loading, considering the fact that a typical tooth experiences a million or so loading cycles annually. In the present study, a systematic investigation is described of the effects of prolonged cyclic loading on human dentin in a simulated physiological environment. *In vitro* stress-life (*S/N*) data are discussed in the context of possible mechanisms of fatigue damage and failure. Stiffness loss data collected *in situ* during these tests are used to calculate crack-growth velocities and the fatigue thresholds, and are presented as plots of the crack-propagation rates (da/dN) as a function of the stress-intensity range (ΔK). The *S/N* and da/dN-ΔK data are discussed in light of a framework for a fracture mechanics-based methodology for the prediction of the fatigue life of human teeth.

Introduction and Background

Dentin lies between the exterior enamel and the soft pulp in the core of human teeth. It forms the bulk of the interior structure of the tooth and hence is critical to its structural integrity. Consequently, it is vital that the mechanical properties of dentin are known so that realistic predictions can be made for the effect of microstructural modifications due to caries, sclerosis, aging and dental restorative processes on tooth strength. While several studies have focused on evaluating these properties [e.g., 1-6], there is little consistency in the available data. Exposed root surfaces in teeth often exhibit non-carious notches in the dentin just below the enamel-cementum junction; the etiology for such lesions is believed to involve a combination of erosion, abrasion and abfraction. These notches serve as very effective stress raisers and hence, can be sites for failure due to fracture. Although such fractures have not been studied extensively, it is generally believed that tooth failure is associated either with catastrophic events induced by very high occlusal stresses or, more plausibly, by cyclic fatigue-induced subcritical crack growth. In view of this, it is surprising that so few studies have investigated the effect of prolonged fatigue cycling on human dentin, particularly as such information is also important for the development of replacement materials for restorative dentistry.

In engineering terms, fatigue is used to describe the response of a material to repeated application of stress (or strain). Obviously, the prediction of a time to failure under such loading is critical for engineering design and durability assurance. The classical approach to fatigue, often called the “stress-life” or “*S/N*” approach, has involved the characterization of the total life to failure in terms of a cyclic stress range. This approach involves the estimation of the number of such cycles required to induce complete failure of a nominally flaw-free “smooth-bar” specimen at a given alternating (σ_a) and mean (σ_m) stress. The measured fatigue lifetime represents the number of the cycles needed to both initiate and propagate a (dominant) crack to failure. *S/N* curves for many materials (e.g., steels) exhibit a plateau at about 10^6-10^7 fatigue cycles, termed the fatigue limit, below which failure does not occur [7]. In the absence of a fatigue limit, a fatigue endurance strength is usually defined as the alternating stress needed to give a specific number of cycles to failure. Both terms are used for traditional fatigue design and life prediction, after adjusting for operational variables such as the presence of notches, the environment (e.g., temperature, pH, humidity), the history and spectrum of loading, etc. [7].

In reality, most structures, including human teeth, have an inherent population of flaws. In such cases, the crack initiation life may be non-existent, thus making lifetimes predicted from the *S*/*N* approach highly non-conservative. A more realistic approach here is to consider that the life is the cycles needed to propagate one such flaw to failure. To make such predictions, fracture mechanics (damage-tolerant) methodologies are generally used, where the number of cycles required for an incipient crack to grow subcritically to a critical size, defined by the limit load or fracture toughness [6], is computed from information relating the crack velocity to the mechanical driving force (e.g., the stress-intensity factor).

The objective of the this study is to characterize the stress-life and crack-propagation fatigue behavior of human dentin *in vitro*, i.e., in Hank's Balanced Salt Solution (HBSS), and to use this information as a preliminary basis to develop such lifetime prediction analyses for teeth.

Materials and Experimental Procedures

Recently extracted human molars, sterilized using gamma radiation, were used in this study. Sections, ~1.5-2.0 mm thick, were prepared from the central portion of the crown and the root vertically through the tooth. The typical microstructure of dentin is shown in Fig. 1. Human dentin is a hydrated composite composed of nanocrystalline apatite mineral (~45% by volume), type-I collagen fibrils (~30% by volume) and fluid, and other non-collageneous proteins (~25% by volume). The mineral is distributed in the form of fine crystallites (5 nm thick) in a scaffold created by the collagen fibrils (50-100 nm diameter). The distinctive feature of the "microstructure" of the dentin is a distribution of cylindrical tubules (~1-2 μm diameter) that run from the dentin-enamel junction to the pulp chamber. These tubules are surrounded by a collar of highly mineralized peritubular dentin (~1 μm thick) and are embedded within a matrix of mineralized collagen (intertubular dentin). The mineralized collagen fibrils form a planar felt-like structure oriented perpendicular to the tubules. There is evidence, although somewhat inconclusive, that the orientation of the tubules leads to anisotropic mechanical properties in human dentin [2,3]. The present study, however, is restricted to a single orientation with cracking nominally perpendicular to the long axis of the tubules, i.e., in the plane of the collagen fibrils. This orientation is believed to have the lowest fracture toughness [2,3].

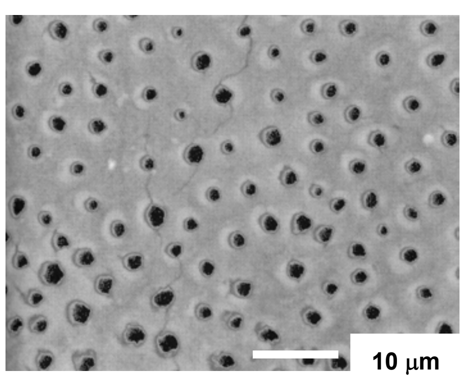

Fig. 1: Micrograph illustrating the typical microstructure of human dentin.

Beams of dentin (0.9 x 0.9 x 10.0 mm) were machined in an attempt to align the tubules to the long axis of the beam. In actuality, it is almost impossible to align the fracture plane precisely with the tubule axes *a priori* because, with the exception of the root, the tubules in dentin do not run a straight course from the enamel to the pulp, but follow a complex, S-shaped curvature [8]. Consequently, the orientation of the crack plane was determined by examination of the fracture surfaces. Samples were obtained from these beams by wet polishing to a 600 grit finish; twenty-five such beams were used in the present study. Each beam included some root dentin and some coronal dentin such that the loading configuration shown in Fig. 2 could be achieved. *In vitro* first yield (σ_y) and maximum flexural (σ_F) strengths levels were measured in HBSS in bending to be $\sigma_y \sim 75$ MPa and $\sigma_F \sim 160$ MPa, respectively. It should be noted here that the "yielding" observed is the result of irrecoverable diffuse damage.

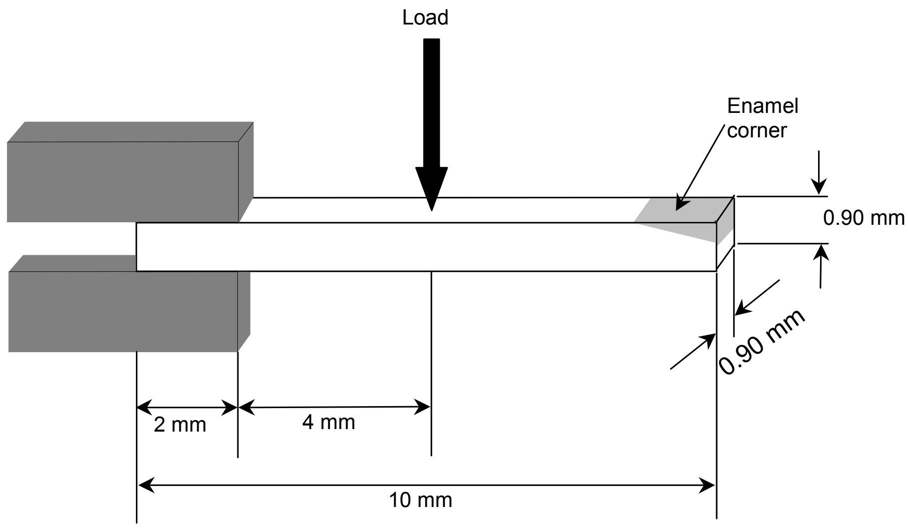

Fig. 2: Schematic illustration of the cantilever beam geometry used for *in vitro* fatigue testing.

In vitro S/N fatigue tests were conducted in ambient temperature HBSS with unnotched cantilever beams cycled on an ELF® 3200 series acoustic testing machine. Tests were performed at a load ratio, *R* (minimum load/maximum load) of 0.1 at cyclic frequencies of 2 and 20 Hz. The beams were cycled to failure under displacement control, with the loads being monitored continuously. Stress-life curves were derived for both frequencies in terms of the stress amplitude, σ_a, (one half of the difference between the maximum and minimum nominal bending stresses). The minimum and maximum stress levels employed ranged between, respectively, ~5 and 135 MPa.

Crack-propagation rates were estimated from the loss in stiffness of the test specimens during the *S/N* tests. Continuous *in situ* monitoring of the specimen stiffness, measured in terms of the bending load and load-line displacement, was used to yield an estimate of the specimen compliance, which was then related to a crack size using standard beam theory and fracture mechanics analyses. The sample compliance was calculated using the analysis for the additional remote-point displacement (rotation), θ_{crack}, due to a crack in a cantilever bending beam [9]:

$$\theta_{crack} = \theta_{total} - \theta_{no\ crack} \ , \tag{1}$$

where θ_{total} is the total rotation and $\theta_{no\ crack}$ the rotation of an uncracked sample, respectively. These rotations can be calculated from standard beam theory, where θ_{crack} is given by [9]:

$$\theta_{crack} = 4\sigma S(a/b)/E' \ , \tag{2}$$

where σ is the maximum nominal bending stress (in the absence of a crack), *a* is the crack length, *b* is the beam width, *E'* is the appropriate Young's modulus, and *S*(*a*/*b*) is given by [9]:

$$S(a/b) = (a/b)^2\{5.93 - 19.69(a/b) + 37.14(a/b)^2 - 35.84(a/b)^3 + 13.12(a/b)^4\}/(1 - (a/b))^2 \ . \tag{3}$$

Through continuous monitoring of the change in compliance, the crack length, a, and hence the crack-growth rate, da/dN, was estimated. The stress-intensity factor, K, can be determined from handbook solutions, in terms of the applied stress range, $\Delta\sigma$, and crack size, a , as [9]:

$$\Delta K = \Delta\sigma (\pi a)^{1/2} f(a/b) , \tag{4}$$

where ΔK is the stress-intensity range ($= K_{max} - K_{min}$), and $f(a/b)$ is a function of the specimen geometry and crack size [9,10], viz.:

$$f(a/b) = 1.122 - 1.40 (a/b) + 7.33 (a/b)^2 - 13.08 (a/b)^3 + 14.0 (a/b)^4 . \tag{5}$$

The crack-propagation results are presented as standard log-log plots of the growth rate per cycle, da/dN, as a function of the applied stress-intensity range, ΔK.

The morphology of the fracture surfaces for both fatigue and overload fracture were examined using a standard scanning electron microscope. Surfaces were sputter-coated with a gold-palladium alloy prior to investigation.

Results and Discussion

Stress-life behavior

Stress-life data, obtained at cyclic frequencies of 2 and 20 Hz with $R = 0.1$, are shown in Fig. 3 in the form of the number of fatigue cycles to failure, N_f, as a function of the applied stress amplitude, σ_a. It is evident that dentin shows "metal-like" fatigue behavior in that fatigue lifetimes increase with decreasing stress amplitudes until an apparent plateau, resembling a fatigue limit, is reached at ~10^6 cycles. Values for this apparent fatigue limit in dentin ranged from 25 MPa at 2 Hz to 45 MPa at 20 Hz, representing values of ~15 to 30% of the measured tensile strength; this is to be compared with most metallic materials where, at this load ratio, the fatigue limit is typically 25 to 30% of the tensile strength.

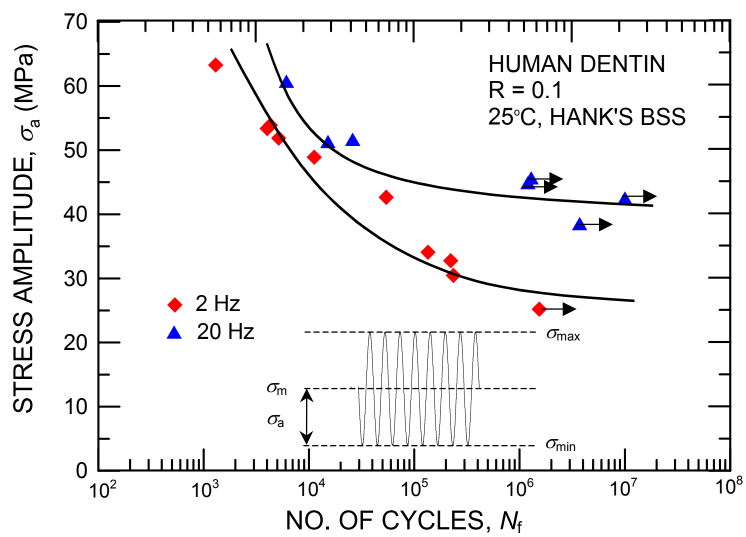

Fig. 3: Stress-life (S/N) data for dentin in HBSS in the form of the stress amplitude, σ_a, as a function of the number of cycles to failure, N_f. Horizontal arrows represent samples that did not fail. Inset shows the definition of the various stresses associated with the fatigue cycle.

Fractographic observations

The fractography of fatigue and overload (fast fracture) failures in human dentin is shown in the scanning electron micrographs in Fig. 4. There was no evidence of fatigue striations, which are indicative of an incremental crack-growth process resulting from alternating plastic blunting and sharpening at the crack tip; this process represents the most common fatigue mechanism in ductile materials [7], but is apparently not operating in dentin. Fracture surfaces representative of cyclic fatigue-crack growth are shown in Fig. 4a. Few differences exist between the morphology of these fractures and those associated with overload failure (Fig. 4b), although macroscopically the overload fracture surfaces are somewhat rougher. Such behavior is more indicative of brittle materials, such as many ceramics and pyrolytic carbon [11], where the mechanisms of crack advance (and hence the fractography) are essentially identical under cyclic and overload conditions; the fatigue effect in these materials is more commonly associated with mechanisms behind the crack tip involving the progressive degradation of the operative crack-tip shielding (extrinsic toughening) processes [12]. There were also no observable differences in the appearance of the fracture surfaces at 2 and 20 Hz.

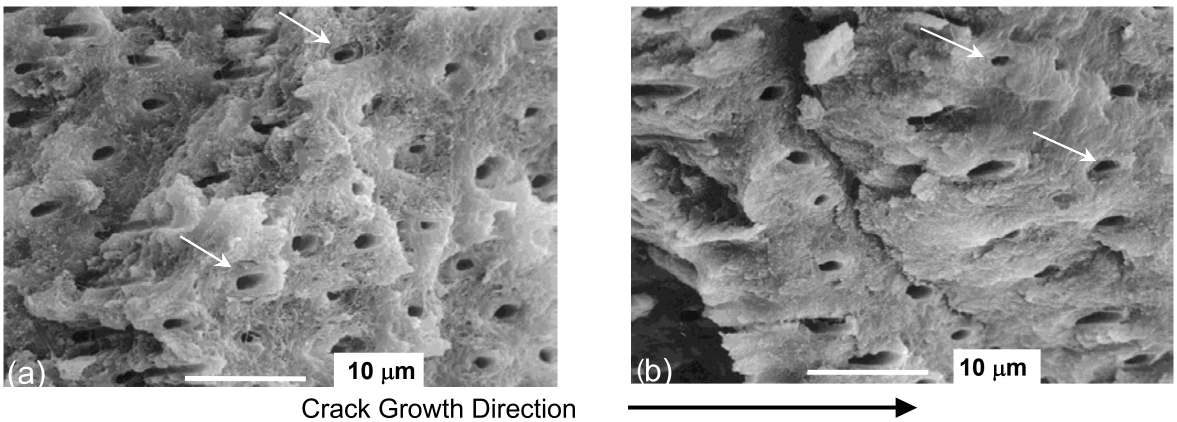

Fig. 4: Scanning electron micrographs of the (a) cyclic fatigue and (b) overload fracture regions. There is some evidence of pullout of the peritubular dentin cuffs (indicated by white arrows).

Fatigue-crack propagation behavior

In the present study, such data were derived from continuous *in situ* measurements of the stiffness loss during *S/N* fatigue tests. Results are shown in Fig. 5; as with many materials [7], the variation in growth rates with ΔK can be described by a Paris power-law relationship:

$$\mathrm{d}a/\mathrm{d}N = C\,(\Delta K)^{m} = 6.24 \times 10^{-11}\,(\Delta K)^{8.76}\ , \tag{7}$$

where C and m are scaling constants (units: MPa√m for ΔK, m/cycle for da/dN). This behavior is consistent with that observed for many brittle materials, which generally have far higher Paris exponents than the m values of ~2 to 4 that are generally reported for ductile (e.g., metallic) materials [12]. The exponent of ~8.76 found for dentin is smaller than that reported for apatite bone mineral substitute (where m ~ 17) [13].

When characterizing the fatigue-crack growth properties of a material, it is often useful to define a threshold stress intensity, ΔK_{TH} (below which crack growth is practically non-existent); this is commonly defined operationally as the stress intensity corresponding to a growth rate of 10^{-10} m/cycle. Extrapolating the current results yields an approximate threshold ΔK_{TH} for human dentin of 1.06 MPa√m. This value is roughly 60% of the measured fracture toughness (K_c ~ 1.8 MPa√m [6]), and is typical of many brittle materials where ΔK_{TH} ~ 0.4-0.6 K_c [12].

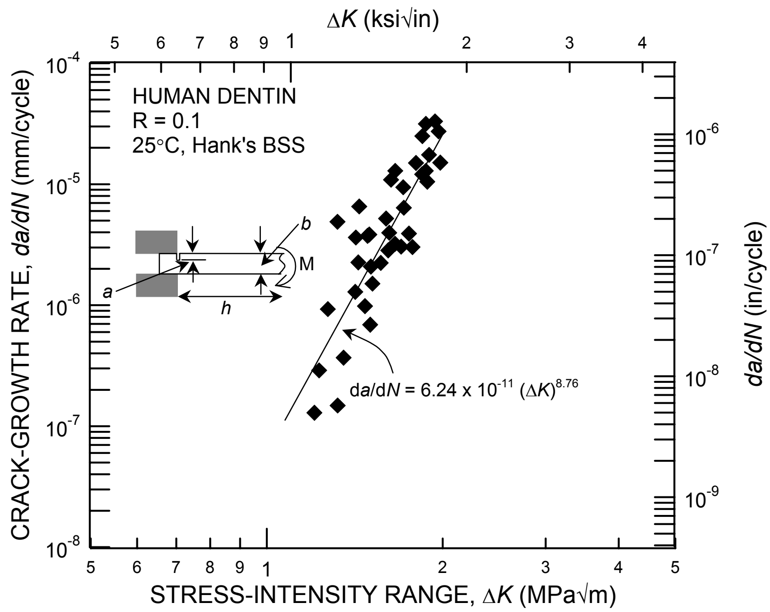

Fig. 5: The variation in fatigue-crack growth rates, d*a*/d*N*, with the stress-intensity range, Δ*K*, for dentin. The inset shows the geometrical configuration used for these calculations.

<u>Lifetime predictions</u>

Typical masticatory stress levels that a human tooth experiences have been estimated to be on the order of 20 MPa [e.g., 14]. At these levels, a fatigue threshold of ~1 MPa√m implies that any cracks formed in teeth would need to be of dimensions in the hundreds of micrometers before meaningful subcritical crack growth can occur. With the availability of the growth-rate data, it is possible to make a preliminary attempt to develop conservative estimates of the expected fatigue life of teeth containing flaws of specific dimensions. This is achieved by integrating the Paris relationship (Eq. 7) between the limits of the initial flaw size, a_o, and the critical (final) flaw size, a_c, for failure; the latter limit is estimated by equating the stress intensity developed ahead of the crack tip to the fracture toughness, K_c:

$$K = f(a/b)\, \sigma_{app}(\pi a_c)^{1/2} = K_c \ , \tag{8}$$

where σ_{app} is the in-service stress, $f(a/b)$ is a function dependent upon the geometry, flaw size and shape, and K_c is ~1.8 MPa√m [6]. The number of loading cycles to cause failure is thus a strong function of the in-service stress and initial flaw size, and can be expressed (for $m \neq 2$) as:

$$N_f = \frac{2}{(m-2)C} \left(f(a/b)\Delta\sigma_{app}\right)^{-m} \pi^{-m/2} \left[a_o^{1-m/2} - a_c^{1-m/2}\right] \ , \tag{9}$$

thereby providing a conservative basis for the prediction of the life of a tooth. As an illustration of this approach, we can assume the presence of a semi-elliptical surface flaw small compared to the dimensions of the tooth. For shallow cracks in this configuration, $f(a/b) = 1.12/\sqrt{\Phi}$, where Φ is the so-called shape factor. Since the in-service stresses are generally small, a worst-case value of 1 can be taken for Φ. Using this configuration for Eq. 9, for an initial flaw of ~100 μm, the predicted lifetime (Fig. 6) at a stress of 20 MPa is well over a billion cycles; at a million cycles per year, this would imply an infinite life for the tooth. However, for a larger flaw of 600 μm, the projected life drops to ~3.6 million cycles, or 3 to 4 years; for a 900 μm flaw, the projected life is as low as a few months. One implication is that any dental reconstruction process that leaves, for example, a millimeter-sized flaw, would mean that the residual lifetime of the tooth could be as short as several weeks, thus necessitating immediate repair.

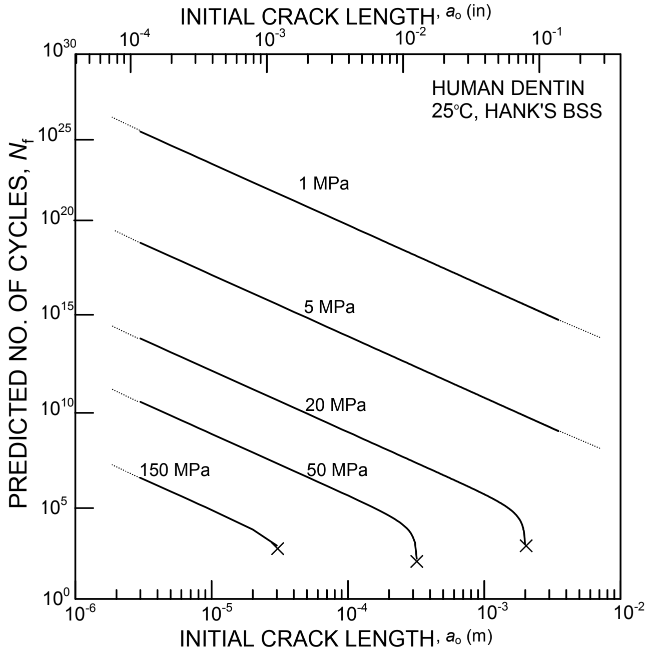

Fig. 6: Predicted fatigue lives, in terms of the number of loading cycles N_f, as a function of the initial flaw size, a_o, for a range of in-service stresses.

It should be noted that this simple analysis is presented as an illustration of how life prediction could be performed for human teeth. Precise calculations would need to consider specific configurations of flaws and better estimates of the *in vivo* stresses and stress-states. However, this approach is inherently more reliable than the traditional stress-life approach, which would not have predicted any failures for physiological stresses of 20 MPa, which is below the relevant endurance strength (Fig. 3). However, several factors that may affect fatigue behavior have not been considered. These include the magnitude and physiological loading history of the tooth, the presence of occasional "over-stresses" [14], the presence of mixed-mode (tensile/compressive plus shear) loads, the physical nature of the flaws, the effect of orientation (if any) on the crack-growth rates, and the occurrence of possible long-term environmental effects on fatigue due to corrosive and bacterial fluids inside the mouth. Hence, though these predictions must be considered as a rough indication of the life of the tooth, they do indicate the general trend that for typical physiological stresses of 5 to 20 MPa, small flaws in teeth of the order of 250 μm will not radically affect their structural integrity, as predicted fatigue lifetimes will exceed the life of the patient. Moreover, this lifetime modeling approach does provide a target flaw size for developing noninvasive means for assessing the risk of tooth failure.

Conclusions

Based on the *in vitro* study of the fatigue properties of human dentin, involving both stress/life and crack-propagation, the following conclusions can be drawn:

1. The susceptibility of human dentin to premature *in vitro* failure by fatigue has been demonstrated, in the form of "smooth-bar" stress-life (*S/N*) curves. These curves (at a load ratio of R = 0.1) displayed a fatigue limit for dentin of ~25 and 45 MPa, for cyclic frequencies of 2 and 20 Hz, respectively.
2. Akin to many brittle materials, the morphology of the fracture surface created during fatigue-crack propagation was identical to that created during catastrophic fracture.
3. da/dN vs. ΔK data obtained suggested a Paris power-law relationship, $da/dN \propto \Delta K^m$, with $m \sim 8.76$. Extrapolation to $\sim 10^{-10}$ m/cycle yielded a fatigue threshold of $\Delta K_{TH} \sim 1.06$ MPa$\sqrt{\text{m}}$.

4. A framework for a fracture-mechanics based life-prediction methodology for the fatigue life of teeth has been developed, and projected lifetimes as a function of the size of pre-existing flaws presented. Based on this preliminary analysis, it is concluded that small flaws in teeth of the order of 250 μm will not radically affect their structural integrity.

Acknowledgments

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